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Single-shot dual-energy x-ray imaging with a flat-panel sandwich detector for preclinical imaging

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We describe a multi-layer ("sandwich") configuration detector consisting of two x-ray imaging flat-panel detectors for single-shot (single-kV) dual-energy imaging. An intermediate copper filter is used to increase spectral separation between the two detectors to improve contrast at the expense of image noise. Monte Carlo and cascaded-systems analyses of the signal and noise performance are described that quantify performance characteristics. Image quality of dual-energy images obtained from a prototype sandwich-detector system is evaluated using a figure of merit (FOM), defined as the squared contrast-to-noise ratio normalized by x-ray exposure for a mouse phantom for preclinical imaging. Demonstration dual-energy bone and soft-tissues images of a postmortem mouse are obtained using the prototype system. While the FOM with the single-shot detector is lower than that achieved using a conventional dual-shot (dual-kV) method, the single-shot approach may be preferable when imaging speed or insensitivity to motion artifacts is a primary concern.

1. Introduction

The projection of three-dimensional (3D) human anatomy on a two-dimensional (2D) projection radiograph results in the superposition of normal tissue, sometimes referred to as "structural noise," that can obscure abnormalities and in some common cases be misread as an abnormality. 3D-imaging methods, including computed tomography (CT) and digital tomosynthesis, are widely used to reduce overlying anatomic structure and improve conspicuity [1].

An alternative approach for reducing background clutter in some imaging tasks is to use energy-discriminating techniques. Dual-energy methods enhance material content (e.g. bone or soft tissue) within a 2D radiograph by combining two (or more) images obtained at different x-ray energies [2]. Receiver-operating characteristic studies have shown that dual-energy methods can improve conspicuity of lesions in particular examinations, such as the detection and characterization of small lung lesions, compared to conventional digital radiography for the same patient dose [3].

Commercial dual-energy imaging systems employing flat-panel detectors (FPDs) currently use a dual-shot approach that acquires low- and high-energy projections in successive x-ray exposures by rapidly switching the kilovoltage (kV) applied to the x-ray tube. However, the time interval between exposures can result in motion artifacts that must be addressed. For example, Shkumat et al. [4] developed a pulse-oximeter-based gating method to minimize motion artifacts in cardiac imaging by restricting both high- and low-energy acquisitions to diastole. For non-gated acquisitions, motion artifacts can restrict the successful application of dual-shot methods to relatively stationary and cooperative patients.

An alternative method is to use a single-shot approach to dual-energy imaging by acquiring two images simultaneously, such as by stacking photostimulable phosphors (PSPs) in a sandwich configuration [5,6]. The front layer absorbs (primarily) low-energy x-ray photons while the rear layer absorbs (primarily) high-energy photons. Early investigators examined the use of different detector systems. Speller et al. [7] used film–screen pairs but found their dynamic range and speed insufficient for practical use. Brooks and Di Chiro [8] and Fenster [9] investigated a xenon split-detector

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design for CT. Barnes et al. [10] demonstrated single-shot images using a pair of scintillator-coupled linear photodiode arrays that required a time-consuming scanning procedure for area images. More recently, Allec et al. [11] reported on the feasibility of using amorphous-selenium layers. Their demonstration single-pixel detector showed good agreement with a theoretical model of signal and noise and suggested practical feasibility.

While single-shot methods are more tolerant of patient motion and less susceptible to motion artifacts, they generally suffer from reduced contrast-to-noise ratio (CNR) compared to dual-shot methods for the same total patient dose due to poor spectral separation [12,13]. Spectral separation can be improved using layers of differing atomic number, but that is not always practical. Ergun et al. [14] showed image quality could be improved using PSPs by increasing the energy separation between front and rear layers and using scatter and beam-hardening corrections. Alvarez [15] described a hybrid single-dual-shot method using fast kV switching with a novel PSP-based sandwich detector in which the sensitivity of the front PSP was modulated by a custom electro-optical system that erased the high-kV signal to achieve greater spectral separation.

The objective of this study was to investigate the use of single-shot methods for preclinical imaging where motion artifacts can be of greater concern than a modest penalty in CNR for a given dose. We present results of a theoretical investigation (analytic and Monte Carlo) into signal and noise considerations for the single-shot method with comparisons with the dual-shot approach. Experimental results using a prototype detector developed in our laboratory show agreement with theory. Demonstration dual-energy bone and soft-tissue images of a postmortem mouse are shown using both methods on the same detector.

2. Materials and methods

2.1. Sandwich detector preparation

A novel sandwich-style single-shot detector was developed by stacking two FPDs [16]. Each FPD consists of a commercially-available terbium-doped gadolinium oxysulfide (Gd_2O_2S:Tb) scintillator optically coupled to a complementary metal-oxide semiconductor (CMOS) matrix-addressed photodiode array (RadEye1™, Teledyne Rad-icon Imaging Corp., Sunnyvale, CA). The front and rear scintillators are Min-R™ 2000 (34 mg cm^{-2}) and Lanex™ Fast (48 mg cm^{-2}) respectively [17]. The rear scintillator is thicker to achieve high quantum efficiency with the higher-energy spectrum. The CMOS sensor has 0.048-mm pixels arranged in a 1024 × 512 format to provide an imaging area of approximately 50 × 25 mm. A thin copper (Cu) sheet was placed between the two FPDs to improve spectral separation as described below. The sandwich detector was installed in a light-tight aluminum (Al) box with a 1-mm thick polycarbonate entrance window. Design specifications are summarized in Table 1. Particular attention is given to the signal and noise implications of x-ray photons that may interact directly in the CMOS sensors with this design.

2.2. Monte Carlo simulations

Monte Carlo (MC) simulations were performed to evaluate optimal filter thickness and exposure kV. X-ray spectra were generated using an in-house MATLAB® routine that implements algorithms published by Tucker et al. [18] for a tungsten (W)-target x-ray tube. The simulation framework is similar to that used in a previous work [19]. A numerical phantom was used to mimic a mouse for the optimization calculations as illustrated in Fig. 1, consisting of four polyurethane (PU, 0.59 g cm^{-3}) disks with a thickness of 3 mm and an Al (2.7 g cm^{-3}) bar with a thickness of 1 mm embedded in 30 mm of polymethyl methacrylate (PMMA, 1.18 g cm^{-3}). The disks and bar mimicked soft tissue and bone, respectively, with the bar overlaid on two disks. A 128 × 128 matrix of 0.16-mm detector elements was simulated, corresponding to a 20.5 × 20.5-mm image.

Random spatial coordinates (x, y) of individual x-ray photons were generated to simulate a uniform Poisson distribution of x-ray quanta incident on the numerical phantom. The photon energy was determined using a random variable having a probability density equal to the normalized x-ray spectral distribution $q_0(E)/\int_0^{kV} q_0(E)\mathrm{d}E$. The probability of transmission through the phantom, $T(E)$, was calculated using tabulated values of x-ray linear attenuation coefficients μ and thickness t for each material j, equal to $e^{-\sum_j \mu_j t_j}$. This was used to generate a binomial random variable that determined whether each photon was transmitted through the phantom. Similarly, the detector quantum efficiency $\alpha(E)$ was calculated and used with a binomial random variable to determine photons that interact in the detector. The detector signals $d^F(x, y)$ and $d^R(x, y)$, corresponding to front and rear layers, were determined by:

Table 1
Summary of system parameters used in this study. The first column represents the various system parameters discussed in the text and shown in Figs. 1 and 2. The second and third columns show the values of the parameters used in the MC simulations and in the dual-energy measurements.

System parameters	Monte Carlo simulations	Experimental measurements
Mouse phantom		
PMMA (background body) thickness and density	30 mm, 1.18 g cm^{-3}	30 mm, 1.18 g cm^{-3}
Al (bone tissue) thickness and density	1 mm, 2.7 g cm^{-3}	1 mm, 2.7 g cm^{-3}
PU (soft tissue) thickness and density	3 mm, 0.59 g cm^{-3}	3 mm, 0.59 g cm^{-3}
Flat-panel sandwich detector		
Front phosphor area density	~34 mg cm^{-2}	~34 mg cm^{-2}
Intermediate Cu filter thickness	0.1–1.0 mm	0.1–0.5 mm
Rear phosphor area density	~48 mg cm^{-2}	~48 mg cm^{-2}
Front/rear photodiode pixel pitch	0.16 mm	0.048 mm
Front/rear photodiode pixel format	128 × 128 pixels	1024 × 512 pixels
Front/rear photodiode Si-layer thickness	0.002 mm	0.002 mm
Front/rear photodiode Si-substrate thickness	0.725 mm	0.725 mm
Other front/rear photodiode substrate thickness	–	Ceramic 1 mm
System operation		
X-ray source-to-detector distance	–	1000 mm
X-ray tube filtration	2.4 mm Al equivalent	2.4 mm Al equivalent
Applied x-ray tube voltage	40–70 kV	40–70 kV
Exposure	Depending on the numbers of added projections and mAs	Depending on the numbers of added projections and mAs